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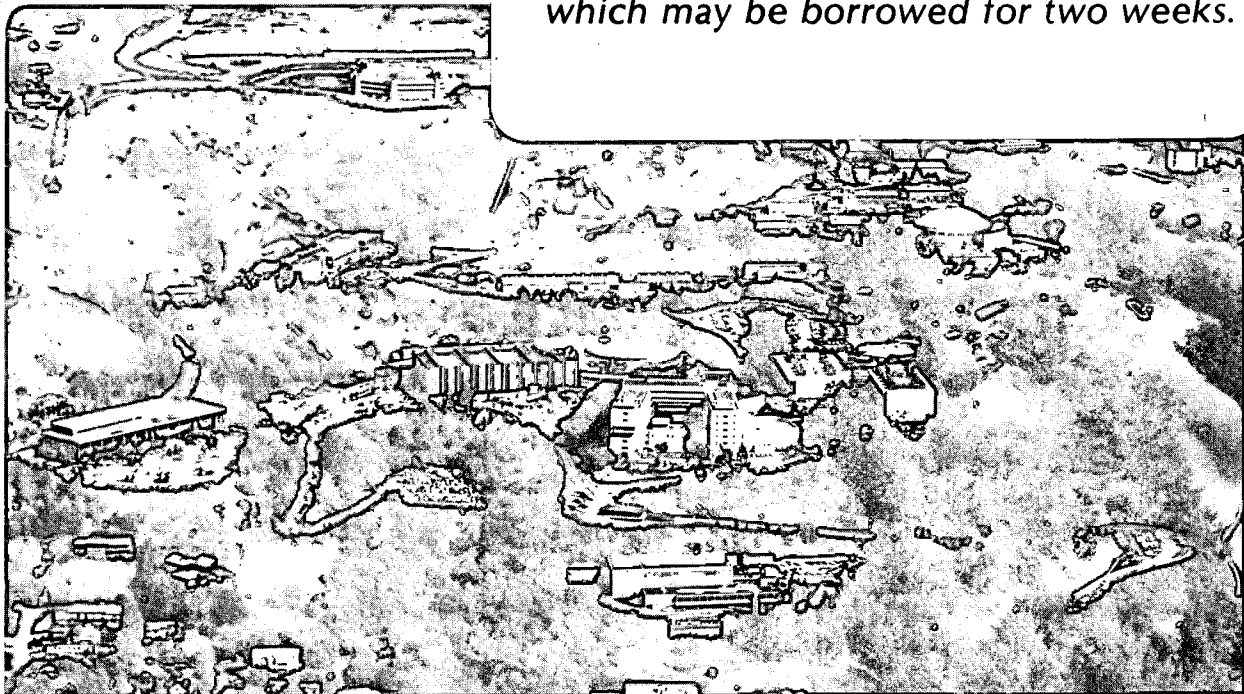# Lawrence Berkeley Laboratory

UNIVERSITY OF CALIFORNIA

Engineering Division

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DISCLAIMER

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ABSTRACT

A field instrument is being developed for the non-invasive determination of absolute lung density using unique Compton backscattering techniques. A system consisting of a monoenergetic gamma-ray beam and a shielded high resolution high-purity-germanium (HPGe) detector in a close-coupled geometry is designed to minimize errors due to multiple scattering and uncontrollable attenuation in the chestwall. Results of studies on system performance with phantoms, the optimization of detectors, and the fabrication of a practical gamma-ray source are presented.

INTRODUCTION

Despite advances in instrumentation for the diagnosis and treatment of cardio-pulmonary disorder, no clinical instrument exists for the accurate determination of pulmonary edema, a condition of excess water in the lung that often results from congestive heart failure and fluid overload when critically ill patients are fed intravenously.

Our objective is to develop a compact, portable instrument for monitoring the course of pulmonary edema at the hospital bedside or outpatient clinics. It must be non-invasive, accurate, sensitive and comparatively inexpensive, and should subject the patient to a much smaller radiation exposure risk than

the standard chest X-ray. By providing fast and direct information on the absolute density of a uniform region of the lung, such an instrument would be extremely useful in the management of patients with a very common and serious condition. It would also prove helpful in clinical research to evaluate the efficacy of therapy.

The difficulty of making an accurate lung density measurement lies in the fact that the lung is surrounded by a chestwall of relatively high density and of variable thickness and composition. Previous attempts to deal with this problem when using gamma-ray techniques are best represented by the so-called "two-source two-detector" method in which the attenuations along the beam paths are supposed to be cancelled when the target density is determined by two sets of transmission and scattering intensities.[1,2] In practice, the method is plagued by a host of problems such as the need to rotate the patient between the two sets of measurements, the difficulty of relocating the target volume and replicating the beam paths. The large size of the apparatus results in limited portability and subjects the patient to a relatively high dose of radiation for adequate counting precision. Above all, multiple scattering is a strong function of system size and mass distribution. Thus the effective target volume can differ significantly from that defined by a pair of collimators, especially where there is little energy discrimination against multiply-scattered photons when detectors exhibiting poor energy resolution are used.

Based on our earlier discovery that the difficulty of the unknown chestwall thickness can be overcome by measuring the relative Compton backscattering intensity per unit distance along the incident gamma-ray beam[3], we are developing a field instrument suitable for conducting _in vivo_ measurements on laboratory animals such as sheep or goats.

METHOD

Figure 1 illustrates a typical experimental setup in which a uniform region of the lung is simulated by blocks of foam plastic of known densities and the chestwall is represented by sheets of plastic placed in the entrance and exit beams. The incident beam makes an angle ϕ with respect to the frontal plane of the detector. In this example, the HPGe detector (35 mm diameter by 17 mm thick) measures the energies of the scattered photons from a collimated beam of incident photons at 166 keV. A lead shield with a recessed rectangular opening (6.4 mm wide by 19.1 mm tall) is placed in front of the detector to define the mean scattering angle Θ and to suppress multiply-scattered photons originating from outside the target volume. A typical energy spectrum is shown in Fig. 2, in which six regions corresponding to scattering from six target volumes at 1-cm intervals along the incident beam are identified. A background subtraction is performed by linear interpolation from counts in the regions I and II near each end of the spectrum.

We found that within certain ranges in Θ and ϕ, N (the scattered count rate per unit distance along the incident beam) decreases exponentially with depth X. Thus $\ln(N) = C + KX$. A striking result is that the negative slope K is insensitive to the presence of the chestwall which tends to affect only the absolute count rate. Furthermore, we have also found that within the range of specific gravities from 0.15 to 1.0, the value K is a sensitive and linear function of target density ρ. Therefore, calibration curves such as those shown in Fig. 3 for various system geometries may be used to calculate ρ once K is determined from spectral analysis.

SYSTEM DESIGN AND EVALUATION

Table 1 summarizes some performance characteristics of the system as depicted in Fig. 1. The gamma-ray beam was from a 3 mCi source of Ce-139 which has a very clean line at 166 keV. The beam has an initial diameter of 5 mm and a half-angle divergence of 50 mr. To ascertain the effectiveness of our present method for determining the density of the lung behind a chestwall of unknown composition and thickness, comparisons were made for the same target with and without the placement of absorbers. Measurements were first made without any plastic absorber to determine K_0. Insertion of the 19 mm thick plastic typically reduced the count rates by a factor of two, but the new value K_1 (without any systematic correction) differed very little from K_0. The corresponding differences in density $\Delta\rho$ between the two cases were calculated according to the calibration curves. When the system configuration was changed, as ϕ was increased from 25 to 45°, there were some improvements in accuracy (mostly at lower target densities). These results suggest that there is a sufficiently wide range of system geometries in which the method can be applied.

Upon further investigation, however, we discovered that these excellent results may be deceptively encouraging. During these tests, we placed no limitation on the size of the lung, but, in reality, the beam also strikes the back wall of the lung and this causes complications. To illustrate the problems and some possible solutions, consider the experimental setup as shown in Fig. 4. Here a beam of 159 keV from a Te-123m source is directed through one side of the phantom at an angle ϕ of 35° (the right front of the body faces the beam directly). The mean scattering angle θ and the corresponding Compton-scattering energy E are labelled at successive 1-cm intervals along the incident beam. The phantom lung is simulated by sawdust of known density held in a 15 cm diameter glass dish. The chest wall is represented by 2 cm of

-4-

water between inner and outer glass dishes. Initially, we believed that the scattering from the back wall could be shielded readily just as from the front wall. As a result of the crowding of the scattering angle and the relatively small change in energy with angle for large-angle Compton scattering, the spectrum is more compressed, and the spatial resolution, as determined by detector energy resolution, is poorer at the rear of the target. Near the front wall, the energy separation per cm is typically a few keV. Thus it is quite easy to shield any direct scattering from the front wall. There will always be a certain amount of off-energy counts reaching the detector via a second scatter at the side wall just in front of the detector. This effect can be minimized by choosing a fit region somewhat deeper into the lung. As the beam reaches near the back wall, the change in mean scattered energy per cm along the beam is typically only a few hundred eV. Therefore, if substantial shadowing is required to adequately shield those events originating from the back wall, the portion of the spectrum for those deeper occuring events may be significantly suppressed. Additional allowance must be made for multiple scattering because, although these degraded-energy events generally fall outside the fit region, they can affect those counts in the background region I, resulting in improper background subtraction. The net effect of a narrowly restricted target area is a decrease in measurement sensitivity. For example, the slope of the K vs. ρ calibration curves may be reduced by as much as a factor of two.

DETECTOR DEVELOPMENT

The HPGe detector with its good energy resolution and its excellent efficiency up to a few hundred keV is ideally suited for quantifying Compton scattered events. Figure 5 illustrates some of the detectors that we have employed. We used first a small cubical detector of approximately 9 mm on each side. Small detectors like this are characterized by their good energy

resolution (better than 600 eV at 122 keV), but their relatively high surface-to-volume ratios may have inherent drawbacks. If the surfaces are perfectly neutral, the electric field lines will be parallel between the electrodes. But if these intrinsic surfaces have acquired a charge state of either polarity, the peripheral field lines will be distorted accordingly and result in degraded signals and a change in the effective volume of the detector. One thus needs to be concerned with the long term stability of the detector performance. The Compton-escape events near these surfaces also tend to result in relatively higher spectral background levels. The small planar detector nevertheless has performed adequately for the early part of our feasibility studies.

To avoid these surface-related problems and to improve counting statistics, we have used a large planar detector (35 mm diameter by 17 mm thick). Taking advantage of the fact that the scattered photons of equal energy are distributed along a conical surface of revolution bisected by the horizontal target plane, it is possible to use a detector shape that is taller than it is wide to improve its geometrical acceptance without sacrificing spatial resolution. A disadvantage of such a large planar detector is its larger capacitance; its energy resolution is about 700 eV at 122 keV. During our system optimization studies, it appeared evident that it would be desirable to expand the range of the target region that is rather free from the effects of multiple scattering from the chest wall. This calls for a detector that has better energy resolution and is shaped to accept events from the target region with high efficiency while having a sharp cutoff for those events (usually entering the detector system at larger angles) originating near the front and especially the back chestwalls.

The third detector we used has most of the desired characteristics. As shown in Fig. 5, it is a composite detector with its central region (1 cm wide

by 3 cm tall) surrounded by an active guard-ring. Events which deposit partial energies in both the central and the peripheral guard-ring are rejected by anti-coincidence to reduce spectral background. With this design, precise detector geometry can be maintained indefinitely because the detector volume is defined by parallel field lines between the central and guard-ring regions free of external surfaces. Unfortunately, we found that the system was saturated with anti-coincidence and electronic cross-talk when it was placed in close proximity to gamma-ray sources that are, unfortunately, contaminated with high-energy lines that can penetrate the 2 cm thick tungsten wall of the source holder.

We are presently fabricating a new detector that incorporates all the desirable features mentioned above. From the detailed illustration in Fig. 6 it can be seen that this shaped-field design resembles one half of a coaxial detector. The boron-implanted surface forms a wrapped-around anode that faces the incoming radiation as shown in Fig. 4. Since most interactions occur near the front (p^+) portion of the detector, the intrinsic surface is tucked behind the detector where any change in the field lines due to the surface state is inconsequential. To maximize the detector energy resolution, the capacitance is reduced by forming a narrow strip of lithium-diffused cathode. We hope that this new design will enhance our ability to better cope with the problem of multiple scattering while capitalizing on the method of measuring the relative scattering in the target lung which is insensitive to attenuation in the chestwall.

GAMMA-RAY SOURCES

Another crucial part of the instrument is the gamma-ray source. Due to a multitude of requirements, the search for a suitable source has been a significant challenge. An ideal source must be monoenergetic with an energy between

100-200 keV. In this energy range, Compton scattering is the dominant interaction process in tissue. Below 100 keV, the angles of Compton scattering are harder to resolve and photoelectric absorption plays a larger role, especially in bone. At much higher energies, the efficiency of germanium detectors decreases rapidly while the shielding problems become increasingly more difficult. Other technical requirements include the absence of neighboring lines where they would interfere in the Compton-scatter spectra, and the absence of higher energy lines which can present serious shielding difficulties as well as raising the spectral background.

Calculations indicate that, in order to make an adequate lung density measurement in one minute (averaging over many breathing cycles), the source strength should approach 1 Ci (37 GBq). Further, the specific activity should be high (preferably much more than 1 Ci/g), so that a near point-source (under a few mm diameter) may be formed without excessive self-absorption. As a practical matter, it should have a reasonable half-life (a few months), and last but not least, it must be feasible for it to be produced economically on a large scale. This last requirement essentially ruled out all accelerator-produced sources in favor of nuclear reactor-produced ones (ie., neutron-capture).

Table 2 lists all the sources that we have used in our studies, along with their decay mode, gamma-rays, half-lives, and means of production. Qualitatively, Ce-139 is the ideal choice. Unfortunately, the natural abundance of Ce-138 is only 0.25% so that highly enriched material is not presently available. The 5 mCi Co-57 which we used in the earlier experiments is prohibitively expensive in large quantities. Besides, there is some interference between its 122 keV and 136 keV lines. Even though its 692 keV line is weak, it can still present a shielding problem at high source strength as well as causing excessive detector background.

The best candidates for practical gamma-ray sources for this purpose are Te-123m and, to a lesser extent, Ce-141 because highly enriched precursor isotopes are available from the Oak Ridge National Laboratory (ORNL). Because of the lack of accurate data on the production and especially burn-up cross-sections of these isotopes via (n, γ) reactions, we have conducted an experiment at the Advanced Test Reactor at the Idaho National Engineering Laboratory (INEL). The very high burn-up cross-section of Te-123m (thousands of barns) limits its specific activity to less than 2 Ci/g at a thermal flux of 2×10^{14} n/cm^2/s. Our attempts to mask the resonance absorption in the samples by enclosing them in Hf foils during neutron irradiation did not produce a positive result. The specific activity for Ce-141 was about a factor of two higher.

Two higher-level sources were fabricated and irradiated at the Brookhaven National Laboratory (BNL) reactor. These used 99.6% isotope-enriched Ce-140 and 96.5% isotope-enriched Te-122 encapsulated in high purity aluminum by electron-beam welding. We found that trace contaminants in the high-purity aluminum used were activated to produce high-energy gamma-rays from Fe-59 and Co-60. The 70 mCi Ce-141 source also contained a large amount of Eu-152 and other high-energy (MeV) beta emitters and their bremsstrahlung radiations without identifiable gamma lines. On the other hand, the 10 mCi Te-123m source possibly contained Zn-65 and Sc-46. The 2 cm thick tungsten shielding was more than adequate to shield the gamma-lines of interest but was so transparent to the high-energy radiation that the detector system was saturated for several weeks after the sources were made.

These results suggest that Te-123m offers the best hope for a practical gamma-ray source. It appears that isotope enrichment of Te-122 by means of ultra-centrifuge may be a much more economical method of separating the parent

isotope for irradiation. Meanwhile, we are developing techniques to encapsulate Te-122 in high purity Supersil quartz in which the maximum mineral impurity is under 0.1 ppm. With the shutdown of the HIFIR reactor at ORNL, we shall try to attain high specific activity by irradiating the samples at BNL at a thermal flux of 8×10^{14} n/cm^2/s.

CONCLUSIONS

We have developed a compact one-source one-detector system that has the potential of becoming a clinical instrument for measuring pulmonary edema. Measuring the relative Compton scattering in the target lung along the incident beam has successfully overcome the problem of chestwall attenuation without the need for making transmission measurements and without requiring compensation for source strength. With the latest shaped-field detector design, we hope to further reduce the effect of multiple scattering while maintaining adequate measurement sensitivity. If we are successful in preparing a high intensity Te-123m source with a tolerable amount of high-energy gamma-ray contamination, we will then be in a position to conduct field tests on animals.

According to plans, a series of experiments will be conducted on anesthetized sheep at the Cardiovascular Research Institute of the University of California at San Francisco. Using the Compton densitometer, we may make continuous _in vivo_ measurements on sheep for the appearance and clearance of pulmonary edema which may be induced by intravenous infusion of Dextran and saline solution. These measurements can be correlated to other measurements such as thoracic lymph flow and the determination of extravascular lung water by thermal-dye dilution techniques as well as the lung's wet-to-dry ratio at the end of the experiment. Other validation tests such as comparison with conventional X-ray, CT and MR techniques will also be performed before conducting more comprehensive tests on human subjects.

ACKNOWLEDGMENTS

REFERENCES

1. E.S. Garnett, C.E. Webber, G. Coates, W.P. Cockshott, C. Nahmias and N. Lassen, "Lung Density: Clinical Method for Quantitation of Pulmonary Congestion and Edema," J. Canadian Med. Assoc., Vol. 116, 133-134 (1977).

2. C.E. Webber and G. Coates, "A Clinical System for the in vivo Measurement of Lung Density," Medical Physics, Vol. 9, No. 4, 473-477 (1982).

3. B.W. Loo, F.S. Goulding and D.S. Simon, "A New Compton Densitometer for Measuring Pulmonary Edema," IEEE Trans. Nucl. Sci., Vol. NS-33, No. 1, 531-536 (1986).

Table 1

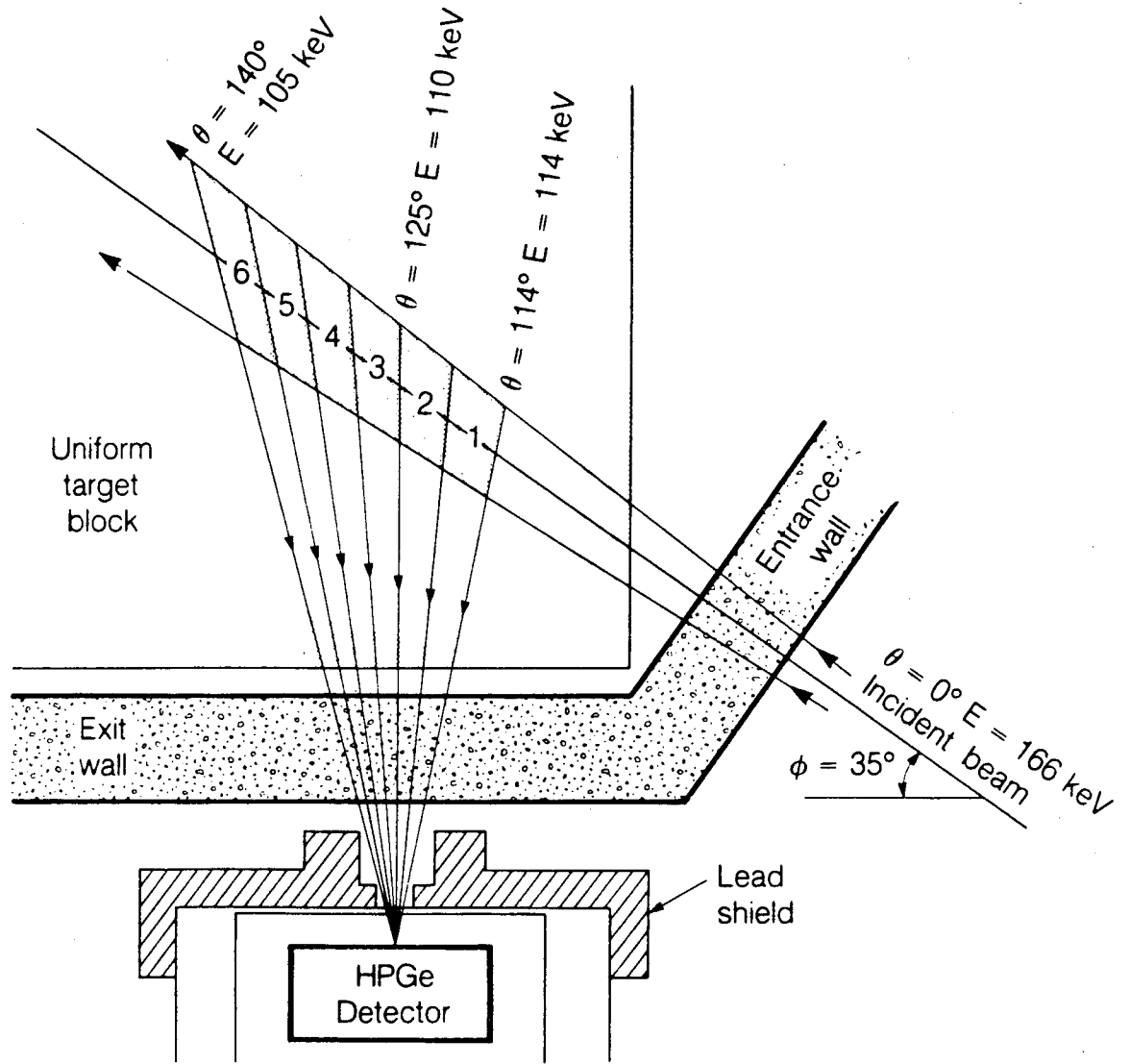| ϕ	ρ(g/ml)→	0.156	0.311	0.644	1.00
25°	K_0	0.134	0.154	0.221	0.296
	K_1	0.147	0.166	0.230	0.299
	K_1/K_0	1.093	1.078	1.041	1.011
	$\Delta\rho$(g/ml)	0.064	0.062	0.046	0.017
35°	K_0	0.153	0.189	0.254	0.325
	K_1	0.161	0.193	0.254	0.330
	K_1/K_0	1.051	1.020	0.999	1.015
	$\Delta\rho$(g/ml)	0.039	0.019	−0.001	0.024
45°	K_0	0.205	0.231	0.299	0.383
	K_1	0.209	0.233	0.297	0.376
	K_1/K_0	1.020	1.006	0.995	0.981
	$\Delta\rho$(g/ml)	0.019	0.007	−0.007	−0.034

Table 1 Test results obtained with experimental setups similar to that depicted in Fig. 1.

Table 2

| Isotope	D.M.	γ (keV)	$T_{1/2}$ (days)	Production
^{57}Co	ϵ	122 (85.5%) 136 (10.7%) 692 (0.16%) Fe X-rays	272	^{56}Fe (d,n) ^{56}Fe (p,γ) ^{58}Ni (γ,p)
^{139}Ce	ϵ	166 La X-rays	138	^{139}La (p,n) ^{139}La (d,2n) ^{138}Ce (n,γ)
^{141}Ce	β^-	145 Pr X-rays	33	^{140}Ce (n,γ)
^{123m}Te	IT	159 (84%) 88 (0.09%) 247 (0.0003%) Te X-rays	120	^{122}Te (n,γ)

Table 2 Some properties of radio-nuclides that have been used in our search for a practical gamma-ray source.

θ = 140°
E = 105 keV

θ = 125° E = 110 keV

θ = 114° E = 114 keV

6
5
4
3
2
1

Uniform
target
block

Entrance
wall

θ = 0° E = 166 keV
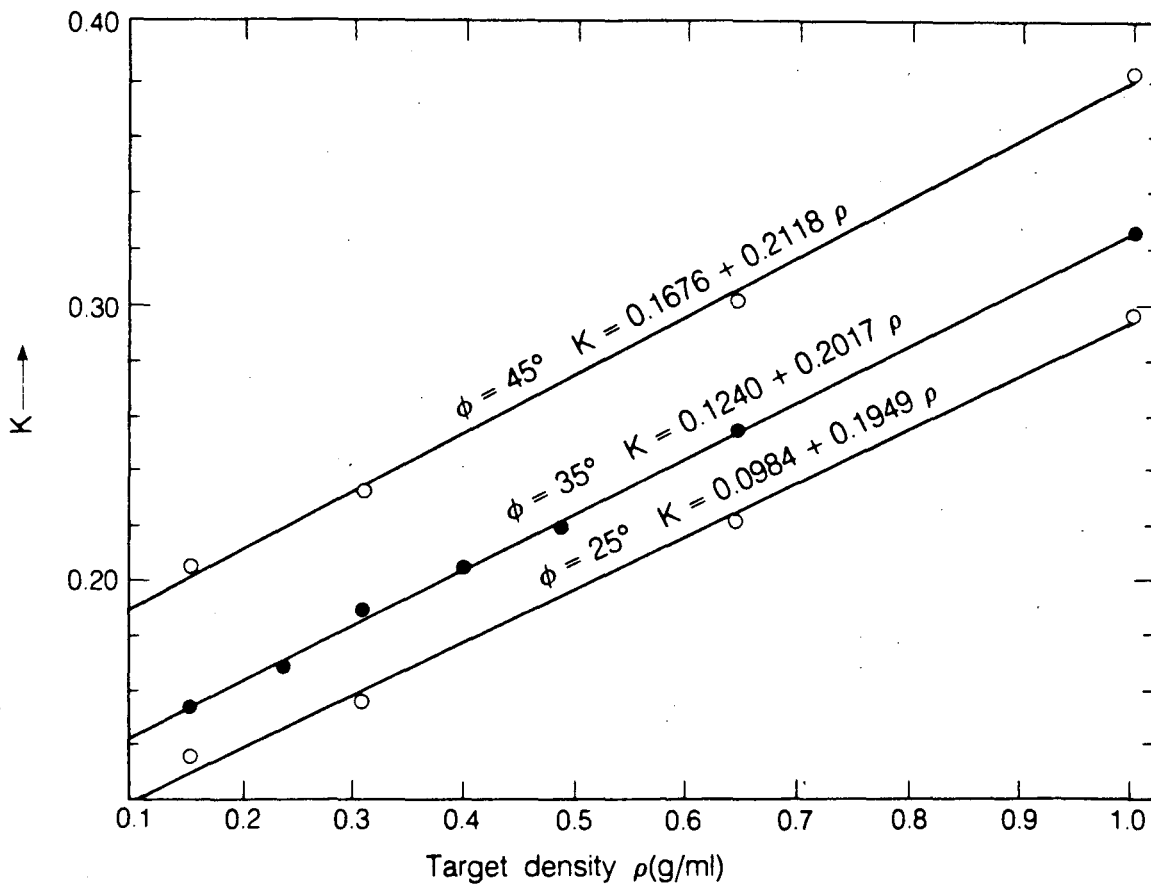Incident beam

φ = 35°

Exit
wall

Lead
shield

HPGe
Detector

XBL 879-3926

Fig. 1   An experimental setup for studying the feasibility of the Relative Compton Scattering technique.
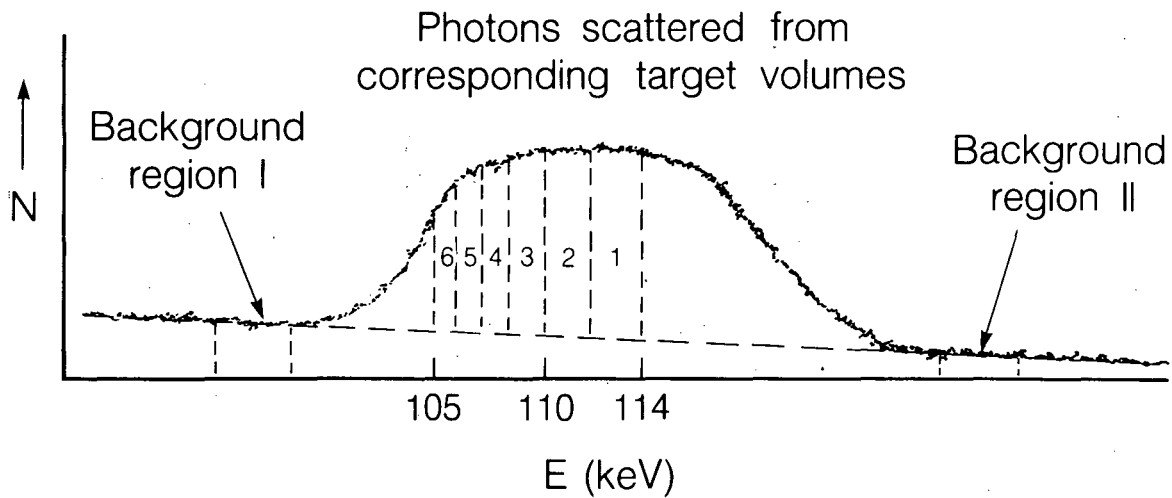
Fig. 2 A typical energy spectrum identifying the fit-regions that correspond to those similarly labeled scattering regions in the target.

Fig. 3   Calibration curves that relate the exponential slope factor K to known target density ρ.
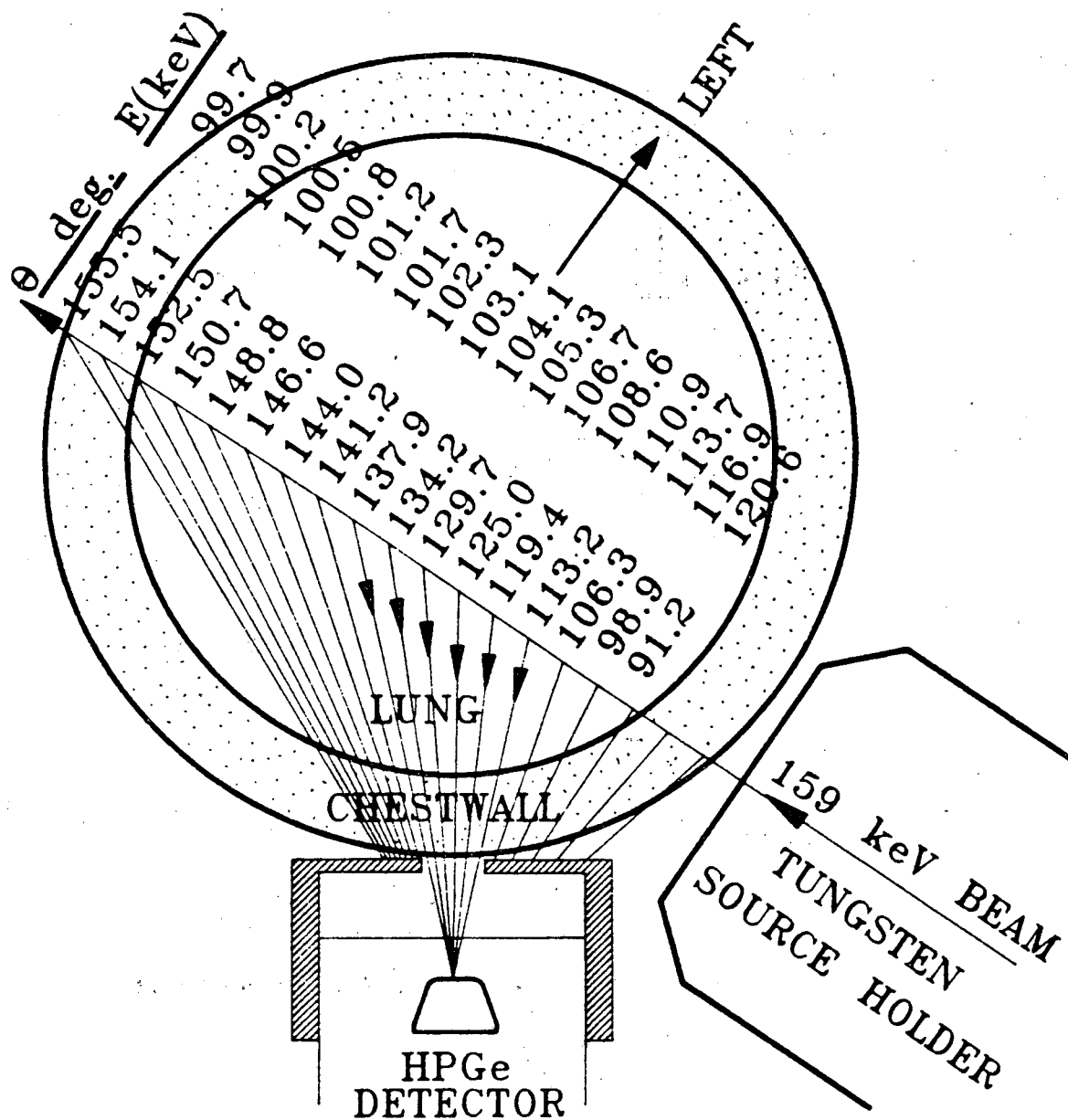
Fig. 4   An improved system geometry to simulate the interaction of the gamma-
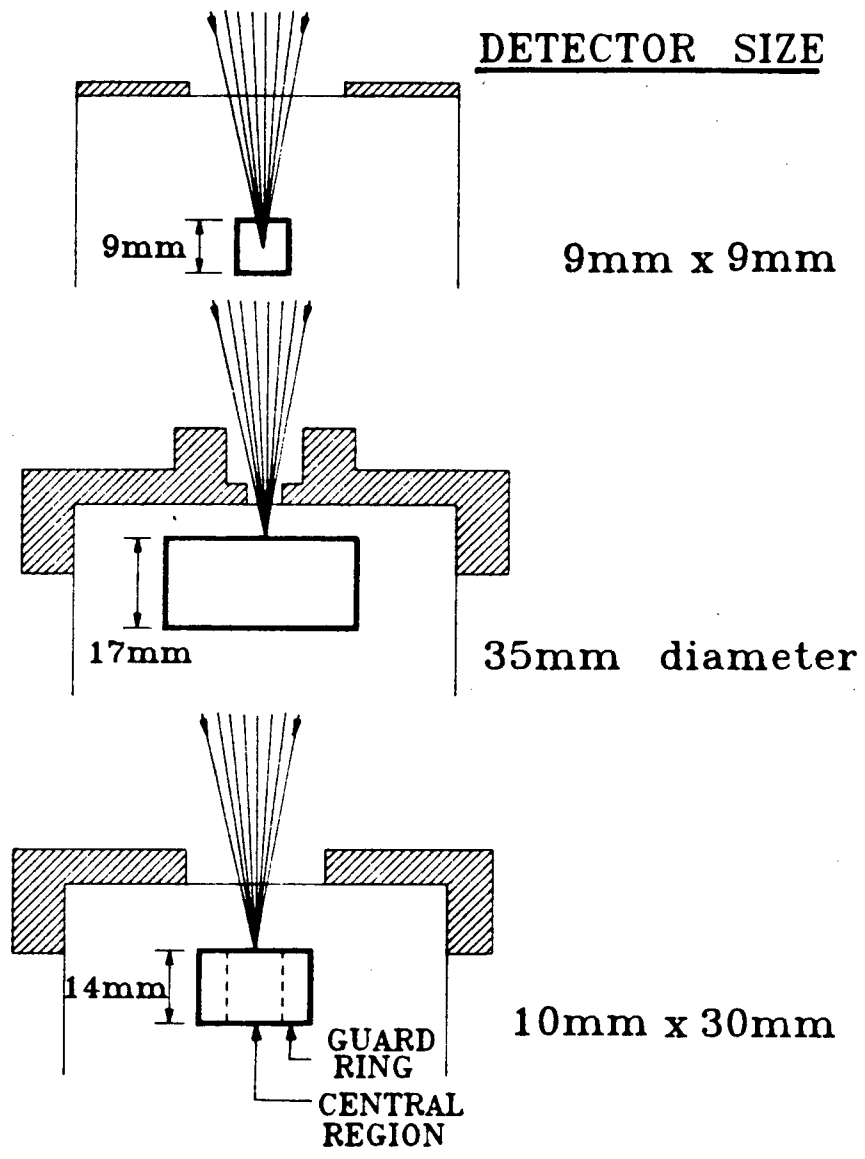ray beam with various points in the chestwall and the target lung.

DETECTOR SIZE

9mm

9mm x 9mm

17mm

35mm diameter

14mm

10mm x 30mm

GUARD
RING

CENTRAL
REGION

Fig. 5   The configurations of several HPGe detectors used in this study.

ANODE
(BORON IMPLANT)

INTRINSIC SURFACE

CATHODE (LITHIUM DIFFUSION)

20mm
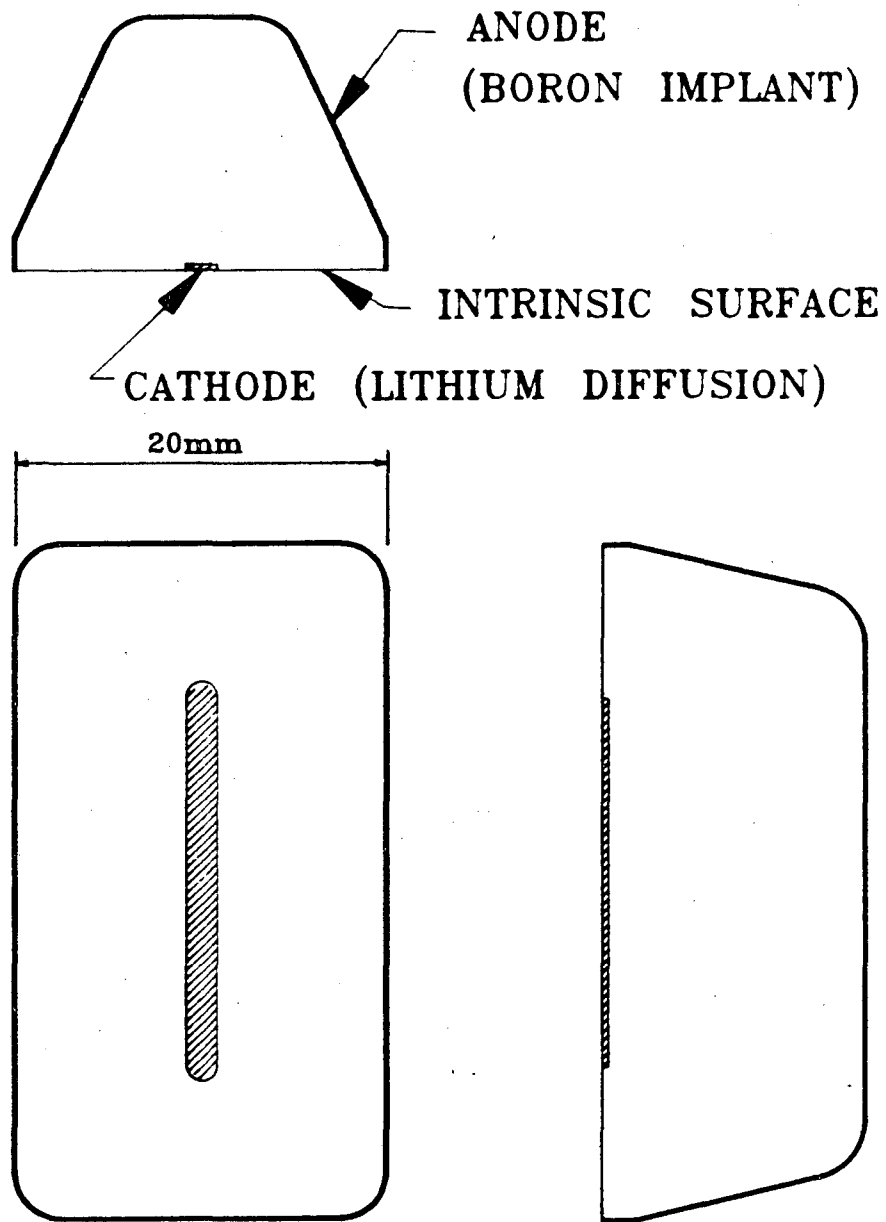
Fig. 6   Details of an optimized shaped-field HPGe detector.